(12) UK Patent Application (19) GB (11) 2 323 207 (13) A

(43) Date of A Publication 16.09.1998

(21) Application No 9704999.3

(22) Date of Filing 11.03.1997

(71) Applicant(s)
Elscint Ltd
(Incorporated in Israel)
Advanced Technology Centre, P O Box 550, Haifa 31004, Israel

(72) Inventor(s)
Alan George Andrew Marcel Armstrong

(74) Agent and/or Address for Service
Marks & Clerk
4220 Nash Court, Oxford Business Park South, OXFORD, OX4 2RU, United Kingdom

(51) INT CL[6]
H01B 7/34 , G01R 33/385

(52) UK CL (Edition P)
H1A AKC
G1N NG38C N571

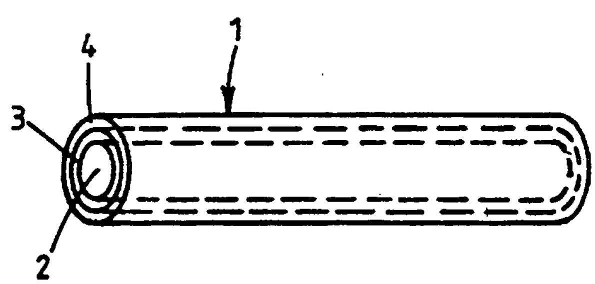
(56) Documents Cited
GB 2040546 A **GB 1105906 A**

(58) Field of Search
UK CL (Edition O) **H1A AKC AKV**
INT CL[6] **G01R 33/385 , H01B 7/34**
On-line: EDOC, WPI

(54) Abstract Title
Flexible hollow electrical cable

(57) A hollow flexible tubular electrical conductor comprises individually-insulated metal wire woven, plaited, knitted or spirally-wrapped into a tubular form which is embedded in and/or impregnated with a flexible electrically-insulating material. The individual insulation of each wire minimises eddy currents. A hollow tube 1 has a passage 2 for water and layers 3, 4 of copper filaments impregnated with polyurethane or rubber. The conductor is used for gradient coils in MRI, an induction furnace or diathermy coils.

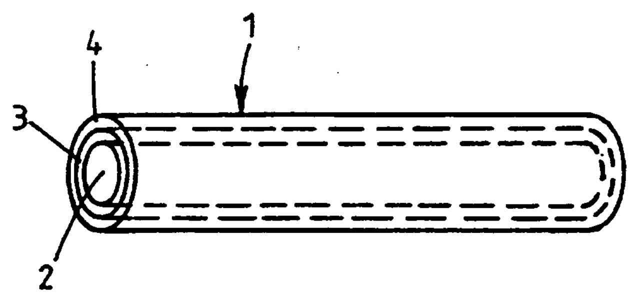
FIG .1



GB 2 323 207 A

At least one drawing originally filed was informal and the print reproduced here is taken from a later filed formal copy.

FIG .1



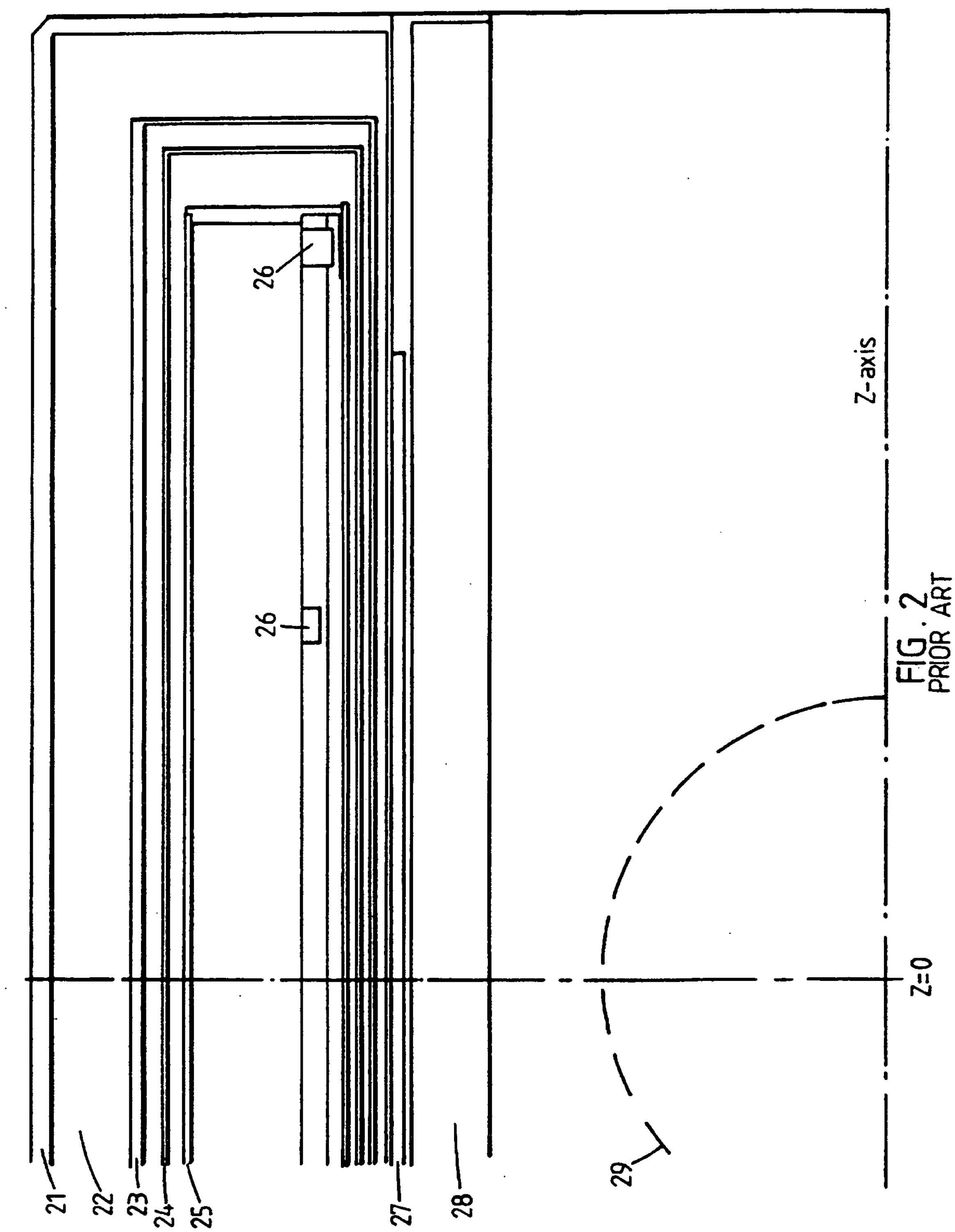


FIG. 2
PRIOR ART

FIG. 3A
PRIOR ART

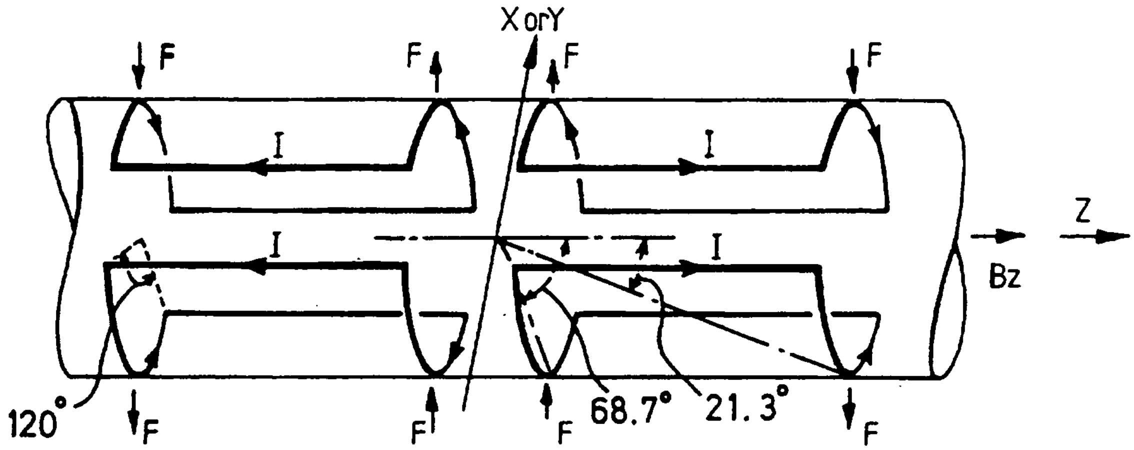
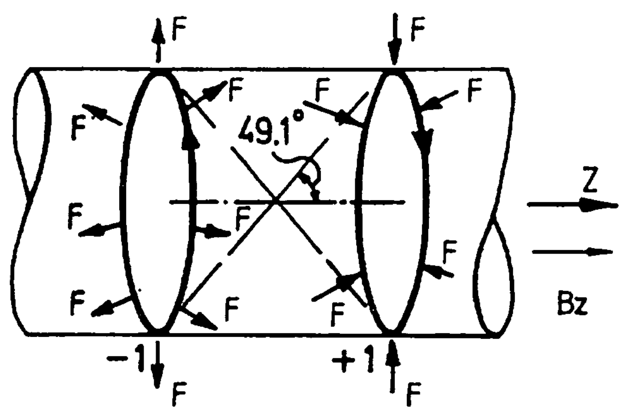


FIG. 3B
PRIOR ART



2323207

M&C Folio: 230P75071 Document #: 70354

Hollow, Flexible, Tubular, Electrical Conductor

This invention relates to a hollow, flexible, tubular electrical conductor which is suitable for, although not exclusively for, use in gradient coils for magnetic resonance imaging systems (MRI).

The present invention has arisen from development work on MRI systems incorporating whole-body gradient coils which can be used either for conventional imaging or more particularly for ultra-fast imaging. Such systems are well known, and need not be discussed in detail here in this specification. The direction of the static magnetic field is called the Z axis, which is the reference direction for the system. The static field is referred to as Bz and the two orthogonal directions are the X and Y axes. There are three sets of gradient coils, X, Y and Z which respectively generate field gradients dBz/dX, dBz/dY and dBz/dZ.

The gradient coils and the radio frequency coils are pulsed on for times ranging from one millisecond to some tens of milliseconds. As is well known, by suitable choices of pulse sequence, pulse length and other parameters, the system is able to reveal a wide range of information about the medical subject, for example to contrast between healthy and diseased tissue, or to highlight blood vessels according to blood velocity.

A number of problems arise from the pulse of the gradient fields. Although the static magnet is a dipole whilst the gradient coils are quadrupoles, they are not orthogonal in an MRI system because of their close proximity: there are many troublesome interactions. Ignoring ferromagnetic hysteretic effects, which principally affect iron-yoke magnets working at lower fields (these are considered in part in European Patent Application 645641), the following description concentrates on superconducting magnet systems which are basically axisymmetric solenoids. With these, the main problems are:

1. Noise caused by mechanical distortion of the gradient coils when the gradient currents interact with the magnetic field of the magnet. This is very disturbing to the subject and to medical staff and can cause fatigue failure of the gradient coils. Noise is in general louder with active-shielded gradient coils, as discussed below.

2. Heat generated in the gradient conductors by the gradient currents. With inadequate cooling, the temperatures can burn the subject, cause the gradient coil structure to fail, and interfere with the magnet shimming which is used for field correction. In the past, it has been usual to use air-blast cooling, but this has several draw backs.

3. Eddy currents induced in the magnet by the gradient coils. These eddy currents decay in various ways and interfere with the MRI images. These effects are intended to be mitigated by the use of active-shielded gradient coils, but some difficulties persist.

With gradient coil conductors made from wide strips of copper (or another conductive material) there are two additional effects which it would be an advantage to remove. Both effects are due to eddy currents in the conductors themselves.

4. The first effect is that there is "cross-talk" between the coils of different axes, for example between the X and Y axes. This is shown by the change of inductance of the Y coils when the X coils are mounted. This cross-talk means that extra power is wasted since there must be dissipation in the coils of the interfering axis. It also means that the settling time of the magnetic field due to a given excitation pulse is lengthened, whilst the parasitic currents decay. A similar effect occurs in the self inductance of a coil with wide conductors.

5. The second effect is that the linearity of the magnetic field gradient changes during the decay time of the parasitic currents, so that the images will be distorted with the evolution of time.

While these two latter effects are smaller that the first three effects described above it is nether-the-less desirable to eliminate them.

An axisymmetrical MRI system is illustrated for example in our British Patent Application 2295020. As shown in Figure 1 of that specification, the subject lies on a straight bed which slides into a tubular tunnel along the system axis (the Z axis). He is surrounded by the radio frequency coils. The looks cover of the magnetic system in turn surrounds the radio frequency coils, and part of the looks cover usually includes a cylindrical copper foil or mesh which is part of the radio frequency system. The gradient coils, surrounded by the magnet, in turn enclose the radio frequency coils.

With reference now to Figure 2 of the drawings of the present application, which is a quarter section of an axisymmetric magnet, the superconducting magnet consists of some solenoid coils 26 of superconducting wire carrying a persistent current. They are held at a low temperature, particularly 4.2K for a NbTi conductor, in a cryostat whose general shape is a short, thick hollow tube. There are four main parts to the cryostat: an outer vacuum chamber 21 which is at room temperature, 300K; an evacuated space 22 separates this from an outer radiation shield 23 which is at 77K; within this there is an inner radiation shield 24 at 20K; and within this there is a helium tank 25 at 4.2K which with the superconducting coils is at room pressure. Figure 2 also shows the passive shims 27, and the position of the active-shielded gradient coils 28. These have a homogeneous (good field) volume 29 as shown.

With reference now to Figures 3A and 3B of the drawings of the present application, the gradient coil system 28 will now be described in greater detail. The gradient coils which suit an axisymmetric magnet are cylindrical in general format. The coils which generate the gradients dBz/dX and dBz/dY are saddle-shaped and effectively constitute two opposing dipoles whose axes are parallel to the X or Y axis, and in both cases are perpendicular to the Z axis. The coils which generate the dBz/dZ gradient are two other axisymmetric dipoles whose axes are with and against the main

magnet. The detailed placement of the conductors is arranged to maximise the good-field volume of the coils and is in general more complex than that shown and than that for simple dipoles. The only parts of the coils which contribute usefully to the gradients are those which carry azimuthal currents. The remaining parts are needed to complete the electrical circuits of the coils.

One known coil arrangement is disclosed in "Resonances in Medicine", Vol 1, pp44-65 (1984) by Romeo & Hoult. This is illustrated in Figures 3A and 3B. Bz represents the magnetic field; I represents the currents in the gradient coils; and F represents the resulting force on the gradient coils.

The Z coils are usually made by winding insulated copper strip into prepared slots in a cylindrical glass reinforced plastic former and filling the residual volume with resin. This makes a tube which can be used to support the X and Y coils. One example of the use of a former, in the manufacture of an MRI magnet with active shielding, is described in our British patent application number 9522968.8 (Publication No. GB-A-   ). In this example, the former has radial flanges which form the side walls of the winding groove, to ensure that all the layers of a coil are wound with the same starting and finishing axial dimensions.

As illustrated in GB-A-2295020 referred to above, the X and Y coils are usually of a "fingerprint" design, each of four coils in a set being made by cutting a spiral slot in a copper sheet, either with a water jet or a punching machine. This makes a fingerprint pattern in the sheet. The sheet, which may be bonded to rubber film, is bent to a half cylinder and is fitted onto the Z coil.

Such coils are expensive to manufacture because the components are hard to make and slow to assemble. At present, copper appears to be the most cost effective conductor, although aluminium could be used instead. One of the purposes of the present invention is to find a quicker way of manufacturing the coils.

The generation of eddy currents will now be described with reference to Figures 2, 3A and 3B. The simple gradient coils of Figure 3A and Figure 3B, when placed into a magnet of the type shown in Figure 2, are closely sleeved in the metal inner tubes of the outer vacuum vessel 21, the outer and inner shields 23, 24 and the helium tank 25. Each tube can carry eddy currents induced when the gradient coils are pulsed on or off.

The strength of the eddy currents depends upon the material and the temperature of the tube, and on amplitude and rise time of the gradient current. Partial compensation for the eddy current is possible over a small volume by adjusting the gradient current waveform, but this method is of limited value because the time variation of the eddy currents is spatially dependent, so that different locations require different compensation in the waveform.

A better method of suppressing eddy currents effects is to use active-shielded gradient coils as referred to above. A secondary gradient set, connected in opposition to, and of larger size than, the primary set, is inserted between the primary set and the magnet. Because of the relative sizes of the various coils, it is possible to cancel most of the gradient field at, for example, the outer shield 23 at 77K, and yet to get a useful non zero gradient field over the homogeneous volume.

There are other considerations which prevent active-shielding from being completely effective: geometric errors in construction and placement of components; eddy currents induced in the radio frequency screen (Figure 2); and the difficulty of designing coils with good shielding and linearity which are not excessively expensive in both construction and operation.

The problem of noise from gradient coils will now be discussed. The main field of the magnet Bz acts on the azimuthal currents shown in Figures 3A and 3B to give the forces F. For the X and Y coils, these forces F principally bend the gradient tube, whilst for the Z coils they tend to give a "Coke bottle" distortion, being radially inward over one region along the axis and radially outward in the corresponding region on the other side of the plane of symmetry. In any case, the forces in a strong magnet can amount to several tons, suddenly applied to the gradient tube. The tube, under such an impact, has many shock-excited modes of vibration which are close to the air-column resonance of the system and can produce noise at 120dB or even more.

In early gradient coils, the conductors were only clamped to the tubes at intervals along their length. There was extra noise because the unsupported conductors were also shock-excited vibrating beams. An extra problem often arose because the copper bars work-hardened with the movement, and broke due to fatigue.

In active-shielded coils, the forces on primary and secondary coils are oppositely directed and the noise is louder if, as is often done, the coils are built on separate tubes.

It is known to glue the tubes together using longitudinal bars, in an attempt to reduce noise. Conventionally, heat has been removed by air-blasting, and this has prevented the tubes from being completely encapsulated.

Other attempts to reduce noise have included the mounting of the conductors in rubber, so that the casing of the gradient set moves more slowly than the conductors themselves as the current is switched on or off. This can reduce the noise by up to 10dB; the long term fatigue life of the conductors is not known. Other attempts to reduce noise include the use of carbon fibre reinforced plastics for the tubes, but this, contrary to expectations, added to the noise level. It might be possible to run the gradient coils in a vacuum, but this would add to the cooling problems and would not avoid transmission of noise by the support structures.

Another problem with such systems has been heat generation and removal. The gradient coils are resistive and usually dissipate several kW of power. Air blast cooling is used to dissipate heat, but inefficiency in heat transfer to air still leaves hot spot temperatures up to 100°C in the gradient structure, requiring the use of special materials. The patient must not be exposed to surfaces exceeding 40°C. The looks cover is used as part of the ducting system, but this does not help much in reducing air-borne noise.

The present invention provides a hollow, flexible, tubular electrical conductor comprising individually-insulated metal wire woven, plaited, knitted or spirally-wrapped into a tubular form which is embedded in and/or impregnated with a flexible electrically-insulating material.

The wire, preferably copper, not only conducts electric current, but also reinforces the conductor, enabling it to be used as a hose for fluid coolant, for example water. The individual insulation of each strand of wire minimises eddy current effects, so that the current distribution amongst various strands is homogeneous. When used with fluid coolant, the proximity of the coolant to the conductor reduces the temperature difference between them and promotes efficient heat transfer.

The use of individual strands in this context arose from a consideration of eddy currents in the fingerprint coils, using wide solid copper strip, described above. Mutual induction effects between the coils of different axes due to eddy currents in the strips suggested that current paths during short pulse sequences differ from those which are followed after the eddy currents have decayed. In other words, the linearity, and the shielding of active shielded coils, has been affected by the wide strips. The present invention arose from this recognition, and from the realisation that a solution might be to use a form of Litz wire in which each strip is sub-divided into parallel, individually insulated filaments which are transposed along the conductor so that all filaments have the same inductance.

Thus the invention includes a coil using such a conductor, and having a fluid cooling system with means for passing coolant fluid axially within the hollow tubular conductor. Pure water often needs to be provided, in any case, for the cryocooler, and so would be available for use as a coolant for the conductors. This has the further advantage that conventional systems would require less modification to introduce the present invention.

Water cooling is also good for noise attenuation. Incidentally, water cooling had been used by Picker International in a concrete-encapsulated gradient set which was very quiet, in 1984, but it has not been usual since then to use water cooling in active shielded gradient coils.

In order that the invention may be better understood, a preferred embodiment will now be described with reference to Figure 1 of the accompanying drawings in which:

Figure 1 is a diagram of a section of a hollow electrical conductor embodying the invention;

Figure 2, to which reference has already been made, is a quarter section of an axisymmetric magnet arrangement for MRI;

Figure 3A is a diagram showing basic gradient coil geometry for the axisymmetric MRI system of Figure 2, showing either the X or the Y coil; and

Figure 3B is a diagram corresponding to Figure 3A, but representing the Z coil.

The electrical conductor of Figure 1 has been developed for use in gradient coils of the type described above. It is cylindrical in this example, although alternative configurations would be possible. The conductor 1 is a hollow tube, defining a cylindrical central passage 2 for the flow of water or other fluid coolant. An inner layer 3 and a superposed outer layer 4 of copper filaments are impregnated with, and embedded in, polyurethane resin or an alternative fixable resin, which is electrically insulating. Each copper filament is separately electrically insulated. The filaments of each layer are woven, plaited, knitted or spirally wrapped, to form a stable tubular structure which resists outward radial forces. In this example, there are two superposed layers, and these superposed layers are spirally counter-wrapped.

The electric conductor 1 is manufactured using conventional cable manufacturing techniques, which need not be described in detail in this specification. For example, the conductor could be manufactured by a co-extrusion method in which the conductive filaments are embedded in the resin during the extrusion process (this would be similar to the manufacture of re-inforced plastic hose).

Instead of the resin, any flexible rubber or other flexible plastics material could be used.

As mentioned above, the separate insulation of the copper wire strands minimises eddy current effects, so that the current distribution amongst various strands is homogeneous. The thickness of the flexible resin or other insulator may be quite small, and this means that the fluid can be in extremely close contact with the conductor metal, reducing the temperature difference between them and promoting efficient heat transfer.

Another advantage is that the conductive strands will strengthen the conductor, and this allows the cooling water to have a higher water pressure, thereby further improving the cooling.

Although the conductor has been described with reference to an MRI system it is not limited to this system. It can be used in any application where reduction of eddy currents, ease of coil winding and ease of water cooling are important. This includes any alternating current system, for example an induction furnace or diathermy coils.

The number and cross-sectional area of the conductive strands will depend on the current to be carried by the conductor.

CLAIMS:

1. A hollow, flexible, tubular electrical conductor comprising individually-insulated metal wire woven, plaited, knitted or spirally-wrapped into a tubular form which is embedded in and/or impregnated with a flexible electrically-insulating material.

2. A conductor according to claim 1, in which the metal wire is of copper or aluminium.

3. A conductor according to claim 1 or 2, in which the electrically-insulating material is a flexible resin.

4. A conductor according to any preceding claim, comprising a multiplicity of individually-insulated strands of the metal wire arranged around the axis of the conductor.

5. A conductor according to any preceding claim, comprising a plurality of superposed coaxial layers of the woven, plaited, knitted or spirally-wrapped metal wire.

6. A conductor according to claim 5, in which adjacent superposed coaxial layers are of spirally counter-wrapped metal wire.

7. A coil for use as a gradient coil in MRI apparatus, at least one of whose windings is of the conductor of any of claims 1-6.

8. A coil according to claim 7, connected to a fluid cooling system comprising means for passing coolant fluid axially within the hollow tubular conductor.

9. A coil according to claim 7 or 8, comprising a coil encapsulant formed of the said electrically-insulating material.

10. A coil according to claim 9, in which the encapsulant is resiliently deformable.

11. A conductor substantially as described herein with reference to Figure 1 of the accompanying drawings.

12. A coil substantially as described herein with reference to the accompanying drawings.

13. A magnetic resonance imaging system, substantially as described herein with reference to the accompanying drawings.